# PRECISION MEASUREMENT FOR MICROSURGICAL INSTRUMENT EVALUATION

Lee F. Hotraphinyo[1], Cameron N. Riviere[2]
[1]Department of Electrical and Computer Engineering and [2]The Robotics Institute
Carnegie Mellon University, Pittsburgh, PA, USA

*Abstract* - **An accurate three-dimensional optical sensing system to track the tip of a microsurgical instrument has been developed for laboratory use. The system is useful for evaluation of microsurgical instrument designs and devices for accuracy enhancement (both robotic devices and active hand-held instrument), as well as for assessment and training of micro-surgeons. It can also be used as a high-precision input interface to micro-surgical simulators. Tracking is done by illuminating the workspace at an infrared wavelength and using optical sensors to find the position of a small reflective ball at the instrument tip. The RMS noise per coordinate is presently 1 micron. Sample results are presented.**
*Keywords* –**Microsurgery, accuracy, optical sensing, tremor**

## I. INTRODUCTION

There is need for high accuracy in microsurgery. One of the most demanding of specialties is vitreoretinal microsurgery and there appears to be a consensus within the field of the need for 10 µm accuracy in tool positioning [1]. Without accuracy-enhancement devices, vitreoretinal microsurgeons are capable of accuracy of roughly 60 µm, and that only for brief periods of time [2]. Microsurgical enhancement devices (robotic manipulators [3,4] and active hand-held instruments [5]) are being developed, but a system is needed to sense an instrument's tip position in three dimensions, and with accuracy to less than 10 µm. In addition to evaluating enhancement devices, this sensing system would also be useful for assessing and training surgeons, as well as comprehensive characterization of erroneous manual motion at the microscale.

Ease of manipulation and avoidance of fatigue are important in microsurgery, and as a result, hand-held instruments are of course designed to be lightweight. In order to keep the sensing apparatus from altering the dynamics of the motion to be tracked, it is desirable to avoid configurations that require physical contact between sensor and instrument, or that require significant added mass to be attached to the instrument.

There are numerous commercial systems that are commonly used in tracking surgical instruments. Some of these include Optotrak (Northern Digital, Waterloo, Can.), the miniBird (Ascension Technology Corp., Burlington, Vt.), and the Fastrack (Polhemus, Colchester, Vt.). These systems offer three-dimensional (3-D) tracking and high accuracy, but not high enough for microsurgical tools, and they require an instrument module to be attached to the instrument, resulting in a change to its dynamic response. The MADSAM system has been developed for evaluation of vitreoretinal surgical instruments and surgeons [6]. It utilizes Hall effect sensing and is accurate to a few microns, but only tracks in 1-D.

In response to the need for high-accuracy 3-D instrument tracking for microsurgical instrument evaluation, surgical simulation, and training and assessment of surgeons, the authors have developed ASAP (Apparatus to Sense Accuracy of Position), a high-precision, non-contact system for 3-D tool tip tracking [7]. This paper presents the second generation of ASAP, featuring improved accuracy and linearity, and faster data sampling.

## II. SYSTEM DEVELOPMENT

ASAP tracks the tip position of a microsurgical instrument in three dimensions. The workspace is illuminated by an OD-669 high-powered infrared light-emitting diode (IRLED) (Opto Diode Corp., Newbury Park, California). The IRLED has a peak emission wavelength of 880 nm and is pulsed at 5 kHz with a 50% duty cycle. A white delrin ball 4.7 mm in diameter is attached to the tip of a hand-held microsurgical instrument. The instrument is painted black and all other surfaces within view of the sensors are covered by light absorbing paper. Infrared light from the IRLED is reflected off the ball to two 2-D position-sensitive detectors (PSDs) facing in orthogonal directions, each one oriented at 45 degrees with respect to the IRLED. This provides sensing in three dimensions, with redundant measurement along the vertical axis. Each PSD is fitted with an absorption filter (over 99% transmission for wavelengths above 780 nm and less than 0.001% transmission below 600 nm) and a lens (10 mm focal length; with multi-layer anti-reflection coating). The sensors and the IRLED are mounted on linear stages.

The PSDs used in the system are DL10 duo-lateral super linear detectors (UDT Sensor Inc., Hawthorne, CA). They have an effective sensing area of 10 x 10 mm and absolute accuracy of 99.9% over 64% of this area. There are 4 electrodes connected to the PSD as well as a common fifth pin that can be connected to ground or biased by several volts. It is advantageous to use PSDs since they provide high position resolution and fast response compared with other detectors, such as charge-coupled devices. Several additional advantages of the DL10 PSD are that the position measurements for the $X$ and $Y$ axes are decoupled, improving resolution and linearity, and that it only senses the "centroid of power density" for a light spot with finite area, simplifying the system so it can be treated as a pin-hole camera.

Incident light produces a photocurrent at each of the four output electrodes of each PSD. Each of the four signals passes through a transimpedance amplifier to convert the

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photocurrent to an amplified voltage signal. The transimpedance amplifier is also used to reverse-bias the PSD, which improves frequency response at the expense of a higher noise level.

The converted voltage signals then passes through a balanced demodulator, the AD630BD (Analog Devices Corp., Norwood, MA), which features high accuracy and thermal stability. The modulation/demodulation scheme provides better noise immunity. The AC (desired) part of the signal is converted to DC, while the DC or low frequency (erroneous) part, which is not modulated, is converted to AC by the demodulator. The output of the demodulator is input to an active 4$^{th}$ order Butterworth lowpass filter with a 1200Hz cutoff and unity DC gain. All unmodulated noise, such as that resulting from reverse biasing of the PSDs, is modulated to high frequency and filtered away.

The sensor data ($x$ and $z_A$ for the first PSD), which represent displacement measured from the center of the sensor to the center of the light spot can be calculated as follows:

$$x = \frac{L}{2}\left(\frac{x_1 - x_2}{x_1 + x_2}\right) \qquad (1)$$

$$z_A = \frac{L}{2}\left(\frac{z_{A1} - z_{A2}}{z_{A1} + z_{A2}}\right) \qquad (2)$$

where $x_1$ and $x_2$ are the signals from the left and right sensor electrodes, $z_{A1}$ and $z_{A2}$ are the signals from the top and bottom electrodes, and $L$ is the length of the PSD (10 mm). The second PSD likewise yields $y$ and $z_B$, respectively.

The sums and differences in (1) and (2) are performed in circuitry as analog computations. The four resulting signals, $x$ and $z_A$ for the first PSD and $y$ and $z_B$ for the other, are then passed through another 4$^{th}$ order lowpass filter. The output is digitized by a 12-bit analog to digital converter and sampled by a computer at a rate of 1500 Hz.

Division and gain correction is done in the software. The software also down-samples the four signals ($x$, $y$, $z_A$, and $z_B$) to 150 Hz and then passes them through a lowpass filter with a 25 Hz 8$^{th}$ order cutoff. This additional filter is used to minimize the noise while allowing capture of the full bandwidth of hand motion, including the nominally 8-12 Hz physiological tremor band.

The system is shown in Fig. 1. Fig. 2 displays a testbed that has been developed using ASAP for realistic simulation of manipulation in vitreoretinal microsurgery, including a sclerotomy to accommodate the surgical instrument.

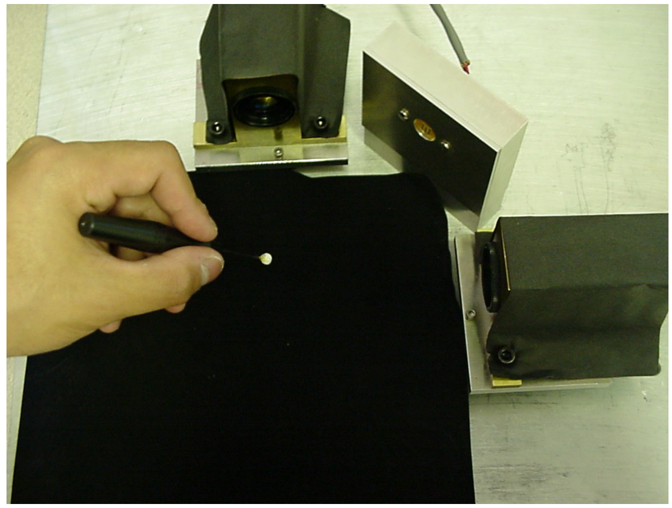

Figure 1. The ASAP system. The apparatus performs 3-D optical tracking of the white ball affixed to the tool tip.

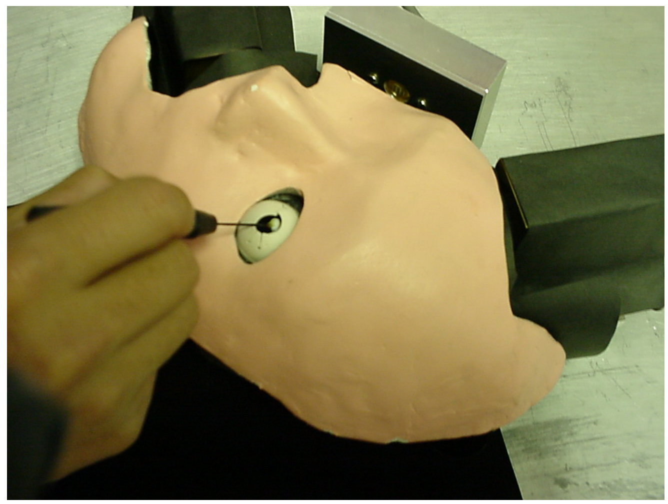

Figure 2. Testbed using ASAP for accuracy assessment in simulated vitreoretinal microsurgery.

## III. System Performance

The RMS noise measured in each of the three axes is approximately 1 µm. Were it not for the final lowpass filter, the RMS noise in each axis would be about 2 µm. The position resolution resulting from discretization is approximately 0.9 µm. With the PSDs having 99.9% accuracy over 64% of their active area, and given the magnification of the lenses, the practical workspace is a roughly cubical region with a volume of about 1 cm$^3$.

Fig. 3 presents a tracing of an orthogonal frame made with an instrument mounted on a manually operated 3-D linear stage. Each line segment was traced twice (once in each direction), showing the repeatability of the system. The figure covers about two-thirds of the available workspace.

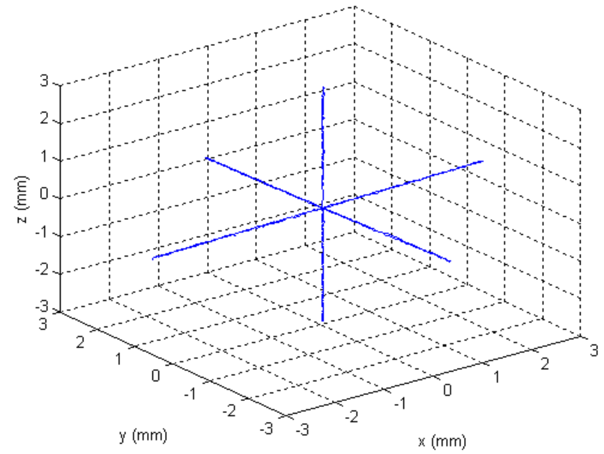


Figure 3. Tracing of an orthogonal frame with instrument mounted on a manual 3-D linear stage.

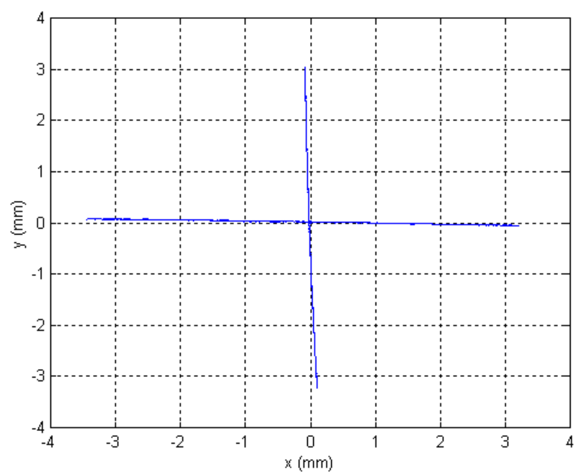


Figure 4. 2-D trace of an orthogonal frame of the *X* and *Y* axis.

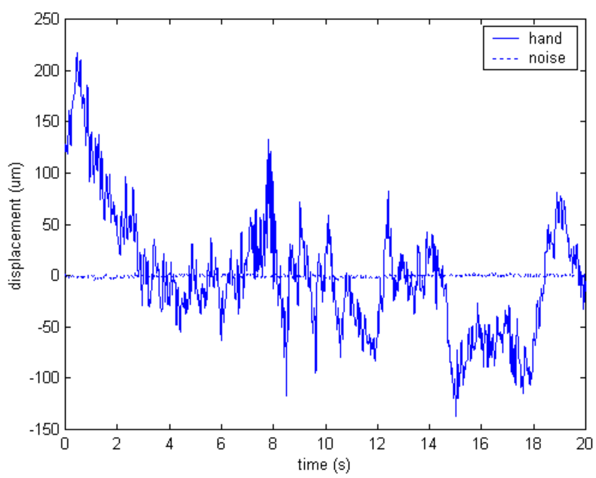


Figure 5. Sample data recorded from subject attempting to hold an instrument motionless in the *Z* axis. The dotted line represents a sample of the noise of the system.

Nonlinear calibration of the system has not yet been performed. Some amount of nonlinearity can be seen in Fig. 4. This is due to misalignment between the linear stage and the sensors, as well as some nonlinearity in the sensors themselves near the edges of the active sensing area.

Figure 5 presents a sample of data recorded while a subject tried to keep a microsurgical instrument motionless. The figure depicts the motion (including physiological tremor and low-frequency aperiodic erroneous drift) as well the noise level of the system, for comparison.

## IV. Discussion

To our knowledge, the data presented here represent the first recordings ever made of microsurgical instrument motion in three dimensions with this degree of accuracy (1 µm rms noise).

This system will be used in experimentation to obtain full 3-D quantification of instrument tip motion by vitreoretinal microsurgeons, as in [8] but with better accuracy. These studies will lead to better understanding and modeling of hand motion components, especially non-tremorous noise components, which are little understood. This information is expected to enable the development of better algorithms for compensating positioning error during microsurgery [9], as well as specification of bandwidth and accuracy requirements for robotic microsurgical accuracy-enhancement device designs. ASAP is also being used for evaluation of such devices [10]. In addition, the system is useful for assessment and training of surgeons, and as an input interface for microsurgery simulators. Non-surgical applications are also possible wherever the marker ball can be affixed to the object to be tracked.

ASAP is a tracking device for the laboratory rather than the operating room itself. A separate research effort in this laboratory has involved the development of an inertial sensing module for microsurgical instruments, which has been used to quantify instrument motion during actual vitreoretinal microsurgery [11].

Nonlinear calibration will be performed to eliminate any non-linearity of the system and to obtain better precision. Future work includes investigation into reducing the size of the reflective ball to improve the usability of the system. This will decrease the signal-to-noise ratio, which may necessitate additional filtering, brighter illumination, or readjustment of the optics. A 16-bit analog-to-digital converter may be implemented in order to increase the resolution of the system.

## V. Conclusion

A laboratory system for 3-D optical tracking of the tip position of microsurgical instruments has been developed using two PSDs to sense light reflected from a small ball affixed to the instrument tip. Noise performance of the system is 1 µm rms in each coordinate direction.


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REFERENCES

[1] S. Charles, "Dexterity enhancement for surgery," Computer Integrated Surgery: Technology and Clinical Applications. MIT Press, pp. 467-471, Cambridge, 1996.

[2] C.N. Riviere, R.S. Rader, P.K. Khosla, "Characteristics of hand motion of eye surgeons", Proc. 19th Annu. Conf. IEEE Eng. Med. Biol. Soc., Chicago, 1997.

[3] P.S. Schenker, E.C. Barlow, C.D Boswell, H. Das, S. Lee, T.R. Ohm, E.D. Palug, G. Rodriguez, and S.T. Charles, "Development of a telemanipulator for dexterity enhanced microsurgery," Proc. 2nd Intl. Symp. Med. Robot. Comput. Assist. Surg., pp. 81-88, 1995.

[4] R. Taylor, P. Jensen, L. Whitcomb, A. Barnes, R. Kumar, D. Stoianovici, P. Gupta, Z. Wang, E de Juan, Jr., L. Kavoussi, "A steady-hand robotic system for microsurgical augmentation," *Lecture Notes in Computer Science*, Springer-Verlag, vol. 1679, pp. 1031-1041, 1999.

[5] C.N. Riviere and P.K. Khosla, "Accuracy in positioning of handheld instruments," Proc. 18th Annu. Conf. IEEE Eng. Med. Biol. Soc., Amsterdam, 1996.

[6] M. U. Humayun, R. S. Rader, A. C. Walsh, C. C. Awh, E. H. Schallen, and J. E. de Juan, Jr., "The objective analysis of vitreoretinal surgical instruments," *Investigative Ophthalmol. Vis. Sci.*, vol. 35, p. 1261, March 1994.

[7] C.N. Riviere and P.K. Khosla, "Microscale Tracking of Surgical Instrument Motion", Proc. 2nd Intl. Conf. on Medical Image Computing and Computer-Assisted Intervention, pp. 1080-1087, Cambridge, England, 19-22 Sept. 1999.

[8] L. Hotraphinyo and C.N. Riviere, "Three-dimensional accuracy assessment of eye surgeons", submitted to 23rd Annu. Conf. IEEE. Med. Biol. Soc., Istanbul, 2001.

[9] W. T. Ang and C. N. Riviere, "Neural network methods for error canceling in human-machine manipulation," submitted to 23rd Annu. Conf. IEEE Eng. Med. Biol. Soc., Istanbul, 2001.

[10] W.T. Ang, C.N. Riviere, and P.K. Khosla, "An Active Hand-held Instrument for Enhanced Microsurgical Accuracy", Medical Image Computing and Computer-Assisted Intervention MICCAI'00, pp. 878-886, Pittsburgh, Oct. 2000.

[11] C.N. Riviere and P. Jensen, "A study of instrument motion in vitreoretinal microsurgery", Proc. 22nd Annu. Conf. IEEE Eng. Med. Biol. Soc., Chicago, 2000.